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Acoustic 3D Imaging of Dental Structures

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Abstract:

Our 3 dimensional elastodynamic imaging project was the first year of a 3 year proposal to combine existing and developing laboratory technologies into a new method of imaging structures which are complicated by high acoustic contrast and lossy / dispersive media. Our ultimate goal is to create a superior technology for 3 dimensional medical imaging.

Our goals for the first year were to perform preliminary investigations and proof of principle tests. This work was to lead to the design of a prototype system for further investigation.

The technologies we brought together were: 1) Individually addressable array construction, 2) Signal pre-processing for propagation enhancement, 3) Signal post-processing for signal and imaging enhancement, 4) Reconstruction techniques, including acoustic tomography for homogeneous media, time reversal scattering for inhomogeneous media characterization, and Acoustic beam forming, 5) computer code modeling of acoustic propagation, through finite difference time domain modeling.

We began work on refining a proposal which would give a new tool to the health professions, and which could be followed up by a simple extension of the idea. We chose the oral cavity because of its accessibility and the enormous range of acoustic impedances encountered there. The five technologies which we applied to this project are: 1) flexible, individually addressable arrays, 2) preprocessing of array source signals, 3) spectral extrapolation of received signals, 4) acoustic tomography codes, including first time of arrival, synthetic aperture time reconstruction, and frequency interpolation and back projection, 5) acoustic propagation modeling code, including finite difference and finite element codes. The computer power needed for the initial stages of this work are readily available.

We based our plan on our ability to make individually addressable acoustic arrays, which can conform to complicated shapes, and preshaped wave forms, which can increase the image contrast and improve its resolution. Our studies for this first year were mainly numerical, and the questions we were striving to answer were 1) how to handle scattering in inhomogeneous media, 2) how to apply modeling and processing codes previously developed to this problem, 3) how to modify developing codes to handle the complications of elastodynamics, and 4) how to deal with saturated, porous media, as in a tooth for example. The answers to these questions are also applicable to other fields of course, most notably shallow, underwater regions and the sea floor among others.

Introduction:

We formed a development plan for 3 dimensional acoustic imaging in difficult elastic media based of the present and projected development of 5 key technologies developed at LLNL. We directed our efforts toward imaging of soft and hard tissue in the human body since 1) this is a useful application of defense technology, and 2) it contains all the most difficult problems we expected to encounter, soft and hard tissue, dispersive media, and difficult geometries.

We chose dental applications because 1) the oral cavity is easily accessible, 2) we hoped to attain much finer resolution than presently available without ionizing radiation methods, CT x-ray for example, and 3) we had access to the experts at U.C. San Francisco's school of Reconstructive Dentistry. We ultimately hope to replace x-ray imaging with a safe, accurate, real-time 3 dimensional imaging system.

Dental structures, in particular the various tissues of the dento-alveolar complex, are particularly suitable for initial use of the proposed technology because of their accessibility, their limited size, and the systematic arrangement of the component tissues. The high elastic contrast and therefore enormous range of acoustic impedances between enamel, dentine, bone, soft tissues and the instruments and materials used in tooth restoration and replacement are also an excellent match with the capabilities of this imaging technology.

There is a major need for improved imaging of teeth and their supporting tissues to enhance the capability for diagnosis and observation of a variety of conditions affecting the teeth and their supporting tissues. There is therefore, we believe, a ready market for the developed technology. Although we focused on dental structures during the grant period because of these factors, the developed technology will also be applicable to other small structures initially and then, as computing power increases, to larger areas of the body.

Conventional, two-dimensional radiographic imaging technologies for dental structures are limited by the nature of teeth and their component tissues, by the physical and geometric requirements of source and sensor positioning and by the challenges. These factors present for the interpretation of resultant images. Root curvature in the long axis of the beam, fracture lines in anything but the beam axis, structures obscured by metal restorations, fissure systems in enamel and remaining dentine thickness in the beam axis are all examples of these limitations.

Three-dimensional radiographic imaging ('CAT scans') might possibly address some of these shortcomings but is limited both by the potential health hazards of ionizing radiation (because of the greatly increased exposure levels required for acquisition of the tomographic data set) and by low image quality. Three-dimensional acoustic imaging now has the potential to overcome all of these problems and shortcomings without health risk.

Acoustic Array Design:

We proposed to decide on a trial array element configuration in our first array. We also proposed to look at the materials available, and likely to soon be available, to design a prototype array for further tests.

We are presently using a PVDF flexible array in a propagation project, and have found that the elements are very sensitive and broad band width with a low Q, but we have also found that the elements lack output power sufficient to use as an imaging device. This material may still suffice, since our present problem is to send signals over very short distances with highly efficient contact coupling.

An interesting material, under design study by Northrop-Grumman and others, is 3-1 composite. This material is an electro-elastic film with piezoelectric stacks embedded within that matrix. This composite is flexible like PVDF, but has very efficient transmission elements built into the controlling film. Though scarce and expensive presently, this material will find wide application in defense applications, and can be expected to become cheaper and more available soon.

Our modeling studies, described below, show that we can create arrays out of either of these materials with element spacing sufficiently close, approximately 1/2 mm center to center separation, to give us the resolution improvement we need to attain 10 um resolution. Addressing the elements individually will require very large computer memory arrays, which we expect will be available by the conclusion of the project.

Signal pre-processing:

Our pre-processing technology was developed in the LLNL Quantitative Nondestructive Evaluation project of 1986-88. This project developed techniques for enhancing the contrast of acoustic measurements by signal pre-processing. This is the technique of inverse filtering which allows us to compensate for the effects of the material's transfer function.

We control completely the shape, and therefore the spectrum, of the initial pulse. Reception of that pulse, acted on by the entire system including the propagation medium, can be by either the sending transducer of another transducer at a different . Our first pulse tells the system how the signal has been changed by traversing the medium between the transmitting element and the receiving element.

If the resultant transfer function is unique, as it nearly always is in linear acoustics, then a unique inverse can be constructed regardless of the system noise. If we then convolve our original, desired, signal with the inverse transfer function and send that signal through the material, then the desired signal, x(t), can be achieved. We then have a way to maintain resolution, or contrast, which will allow us to more clearly resolve the spatial locations of the various interfaces in the media.

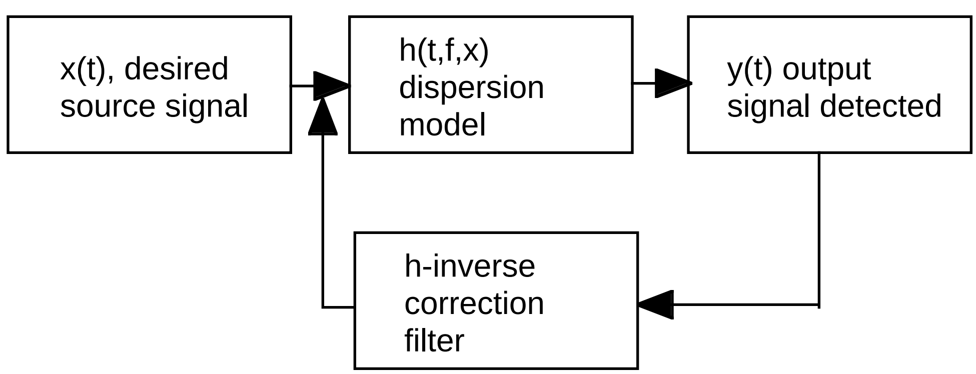

Figure 1. The inverse filter for linear functions. Rather than finding the impulse response of h(t), usually in the Fourier domain, we find the inverse filter correction for h(t) as $F^{-1}\{H(\omega)\} = \{[X(\omega)Y^*(\omega)]/[Y(\omega)Y^*(\omega) + cY^2(\omega)_{max}]\}$.

This can be extended through simple modeling to optimize the resolution inside the material, the interior of a tooth for example, so that we can have location dependent resolution optimization as well. This means that we could send fields which would achieve optimum resolution in any of the parts of the oral cavity structures.

As a test of our technique, we launched pulses of various spectral distributions through saturated solids which modeled the dentine of a tooth or of bone. The received signals were analyzed as to changes made to the initial signal and an inverse filter designed. The reverse filtered signal was then put through the same media and the sharpness of the received signal was attained.

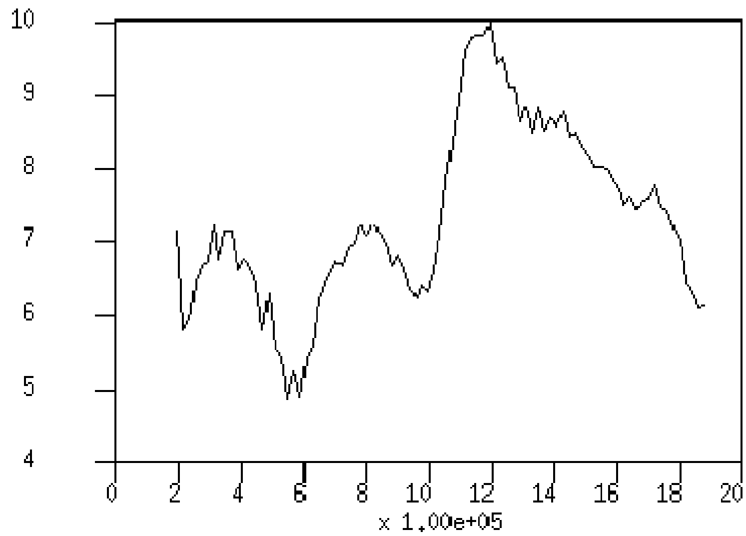

Figure 2. The spectral amplitude for a broad band width acoustic pulse traversing a saturated, porous solid. The maximum variation of the spectrum is 6 dB over a decade of band width, and is easily corrected.

Signal post-processing:

The third technology was also developed in the Quantitative Nondestructive Evaluation project of 1986-88. This project developed techniques for enhancing the contrast of acoustic measurements by signal post-processing. It is based on the mathematical idea of analytical continuation. The technique takes a defined signal and transforms it into the complex frequency domain, which is an equivalent representation. This spectrum is then interpolated from the frequency band of lowest noise to both higher frequencies and lower frequencies. This new spectrum is then inverse transformed and truncated at the same times as the original signal.

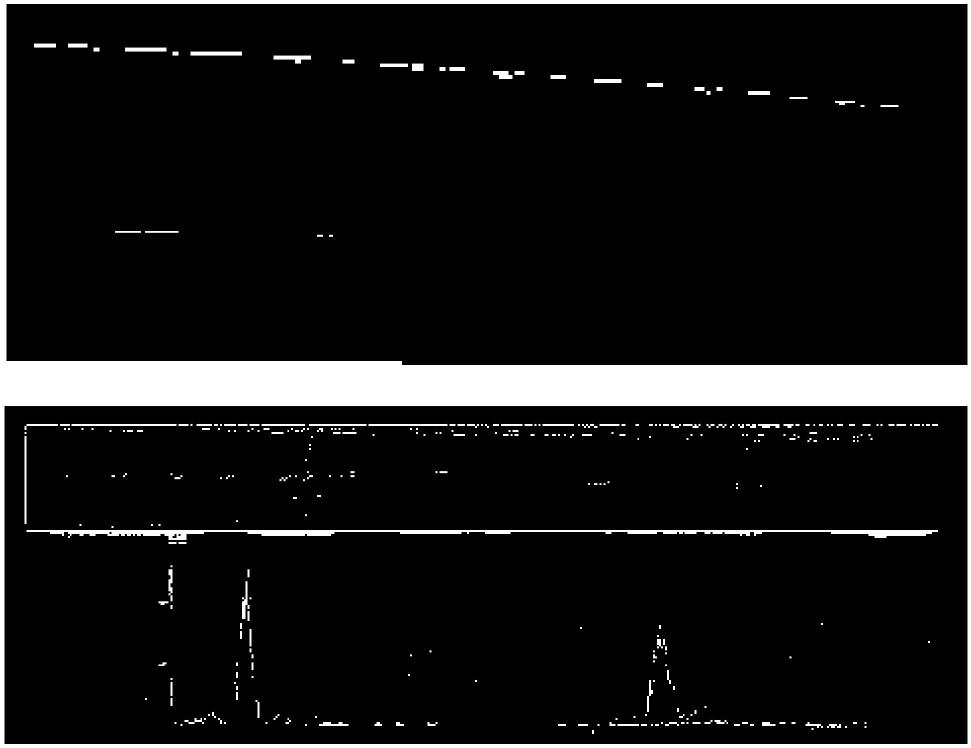

Figure 3. The Figure above is an acoustic scan through a welded plate. The multiple pulse fringes confuse the location of the top and weld interfaces. The bottom picture shows a cleaner picture and a cut along a line reveals that the multiple signals have been replaced by a single marker locating the interfaces. To accomplish this, the spectrum has been extended both towards lower frequencies, to enhance contrast, and towards higher frequencies, to improve sharpness.

In tests performed during the project, the enhancement in the resolution of interfaces was improved by a factor of 10. This signal processing yields the sharpest possible pulse and forms an ideal input to the tomography codes. to the above pulses, we were able to increase the signal to noise ratio by more than 10 dB in all of our tests.

Reconstruction:

Our reconstruction methods were limited to beam forming and time reversal techniques, developed under Navy funding at LLNL. Frequency interpolation and back projection tomography for acoustics was developed and proved in a previous project at Livermore. Beam forming is a model based reconstruction of media using an approximation to the actual situation. Used in an iterative sequence, compared to data and corrected in each cycle, the reconstruction converges rapidly, using a minimum of computer power. The weakness of this approach is the possibility of a bad model, for example an region of inhomogeneous media.

To address this possibility, and provide a means of dealing with it, we investigated time reversal field generation. The basic principle is that in any media, the transfer function for everything, incident wave, scattering center(s) and media, is already in the received field measured by our array. If we simply generate in our second pulse the field received from the preceding pulse, element by element, we will recursively improve the location of the strongest scattering center, suppressing all other scatterers.

Figure 4. The figures, from left to right, show the second radiation of the time reversed field measured after scattering in the volume from the scattering center at the center of the converging waves at far right.

Once we have located the strongest scatterer, we could then remove all trace of it from our initial received field and reconstruct the field we would have received had it not been present. Sending that field back into the media resulted in the enhanced scattering of the second strongest scattering center, and the technique could be repeated for as many locations as needed.

Since our incident field is generated by elements of an array, rather than a single transducer, we could send signals from different angles into the unknown field and determine the relative locations of the same scatterers, as found through different paths in the inhomogeneous media. This results in a faster form of tomography which yields locations and scatterer locations as seen from different aspects.

These successive locations of strong scatterers could then be compared to our model and the locations of known interfaces determined, thus refining the model. Scatterers which do not correspond to know interfaces can be considered as inclusions and further interrogated to determine whether they are caries for example.

The attractive aspects of the time reversal method is its simplicity. The generator used our digital to analog converters for conformity, but the imaging is possible with only first-in, last-out memory. When we removed the strongest scatterer, we needed our full up system, available anyway, but the computations needed were minimal.

To test our reconstruction beam former, we made approximations which let us estimate the accuracy we could expect. We designed a model of pulp surrounded by dentine surrounded by enamel immersed in water. The wave speeds were those found in the medical literature. We based all model wave speeds on the expected wave speed at the location we were beam forming for. We thus used a single wave speed and ignored refraction effects in forming our picture of the model.

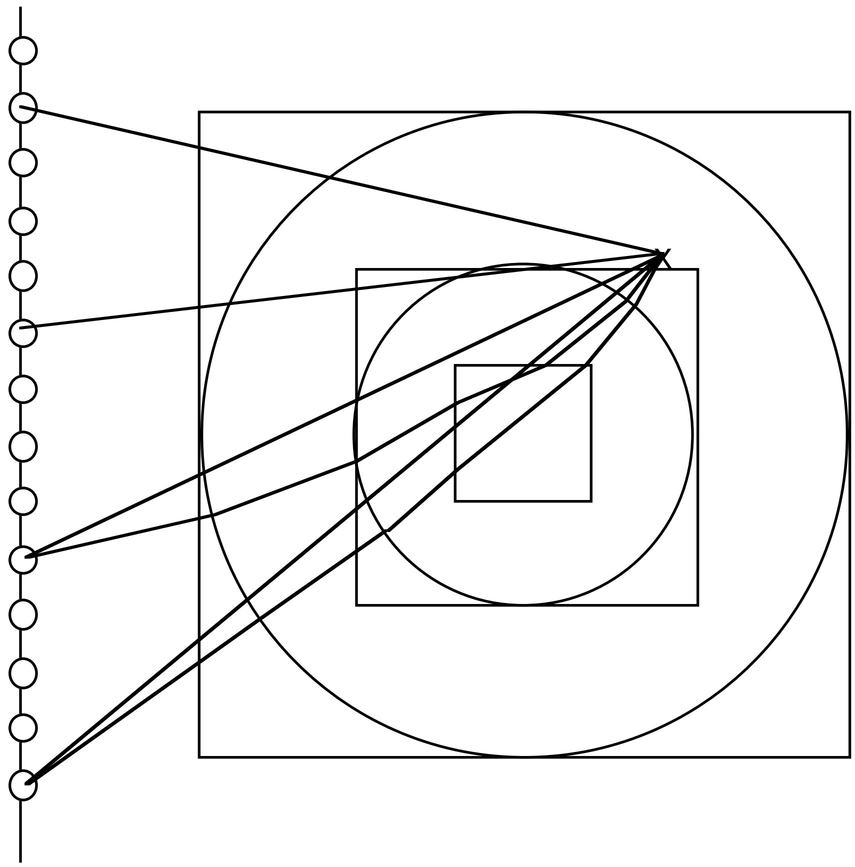

Figure 5. The beam forming problem and our approximations. The actual contours of the 4 region problem are shown in black while our approximated boundaries are red. Furthermore, the actual paths from the beam formed spot (X) to the linear array at left are shown in red while our approximated paths are in black.

We simulated generating fields from different array elements to send energy into the area at different angles and reconstructed the interfaces into a picture. This result was then compared to our model to determine the errors introduced by our original, incorrect, assumptions.

Figure 6. Location of interfaces and scattering centers by approximation beam forming. The figure at the left shows the scattered field from a pulse incident from the left at the center of the array, while the figure at right shows the field from a pulse incident at an angle, launched from -0.5 in array position. Note that the approximations made (see text) yield close to expected results and that the effects of the approximation are apparent.

We found that the fields were as expected in areas where the model matched the actual simulation, but that overall the error, abrupt cutoff of the field at the assumed boundaries, was obvious and could be corrected by minimal changes to the wave speed model. Given that we approached the problem by using incomplete assumptions, in order to minimize computations, our results gave us assurance that we could easily match our model to the reconstructed model.

Acoustic Modeling Codes:

Our original acoustic modeling code was the Acoustic Finite Difference Time Domain (AFDTD) code developed 2 years ago at Livermore. A first generation code, it allowed us to insert a physics package which described elastodynamic propagation including mode conversion from dialational to transverse and the reverse given a preset reflection and transmission coefficient. Its problem, shared by most other similar codes both finite difference and finite element such as Livermore's PING, is that the boundary of a curved body can only be described as a discontinuous line or stair-step contour.

We modeled propagation into a 3 dimensional model of a tooth embedded in bone and surrounded by gingival tissue all immersed in water. We wanted to see if the extremely high contrast of the neighboring tissues would allow propagation throughout the structures and possibly lead to a build-up of heat at the tooth gingiva or bone interface. Our propagation model showed that getting acoustic energy throughout the structures was simple, and that there was no heat build-up at any of the interfaces. We thereby answered the most pressing health question of our project.

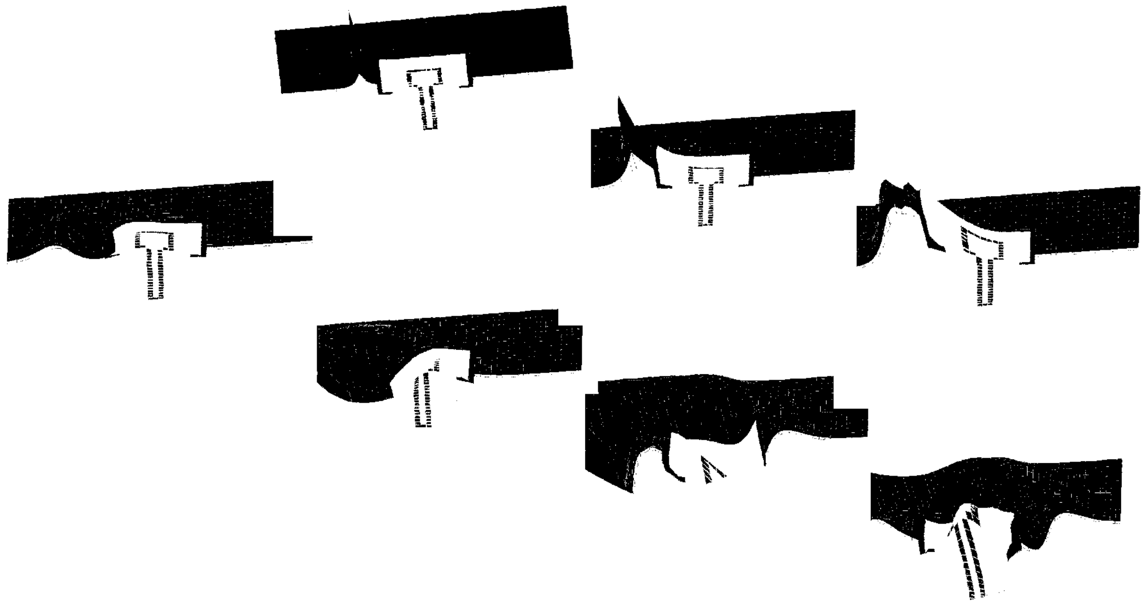

Figure 7. Generation of an acoustic pulse and following penetration of the tooth, with pulp, dentin and enamel regions, bone, supporting the tooth, annd gingiva, surrounding the tooth - bone structure. The initial pulse is generated in fluid. Regions of the problem are represented by color, so acoustic field strength is represented by distortion. Relative field intensity maxima are -39, 0, -8, +0.3, -4, -14, -16 dB, left to right, top then bottom.

Part of our goal was to assist in the development of the DSI-TIGER (Discrete Surface Integral - Time Domain Generalized Excitation and Response) code. The DSI-TIGER code is a new code being built at Livermore to replace many of the time domain electromagnetic codes. Upon completion the code will be able to handle 1D, 2D, 3D structured and/or unstructured meshes. The 3D meshes can be made up of tetrahedrals, hexahedrals, triangular prisms, pentahedrals, or any combination of these elements. The grids can be non-orthogonal thus allowing the mesh to conform almost exactly to arbitrary geometry shapes.

The DSI-TIGER code is being built using modern software design. Specifically, the code is object oriented and is being written using C++. The object oriented approach of DSI-TIGER is very powerful, allowing the physics to be separated from the details of the mesh. This approach allows new physics modules to be plugged into the code with ease, thereby drastically reducing the effort needed to add new physics. We provided the physics of an acoustics package, in the form of elastodynamic equations suitable for a high contrast problem involving Biot solids, and suggested and checked sample problems solved by the code.

This same code will run either on single processor PCs, on workstations or on massively parallel computing platforms. The code also includes the best available radiation boundary conditions required to truncate the problem space.

The DSI part of the code would have allowed us to model curved surfaces by a curve solving for the field at the cell nodes by solving the surface integral describing propagation through and from the curve. Unfortunately, the code was not developed into a useable form by the end of our first year, so results for meaningful problems are unavailable.

To continue our work, we modified the microwave TSAR code to propagate a single type of wave without mode conversion. The results were not fully elastodynamic, but allowed us to check several parts of our project against predictions, notably the time reversal and beam forming work already described.

Summary:

We approached a very difficult problem in a series of steps. Our problem was to find a technique to image in 3 dimensions suitable for use in the human body. To accomplish this, we needed to bring together 5 key technologies in which Lawrence Livermore currently holds the lead. We also needed to make sure that the techniques we used posed no health risks.

Our goal for the first year was to determine how to combine these techniques into a usable method and suggest prototyping a system for further refinement. . The hardware necessary, computation ability and electronics, was assumed to be available by the end of the project.

We investigated flexible, individually addressable acoustic array material to find the best match in power, sensitivity and cost and settled on two candidates. The first, PVDF sheet arrays, has the advantages of being relatively low cost, easily available and already in use in another project. The second, 3-1 composite material, has the advantages of higher sensitivity and power, but lower availability and higher cost. As this material is being applied to many defense related programs, the cost is expected to fall and as the supply rises.

We did not acquire or build any flexible arrays, something we had not proposed to do, but we did acquire data on propagation through saturated, honeycomb structures and hard solids modeling dentine and enamel by using an existing PVDF array. We used published data for the pulp and gingiva since they are essentially fluids.

We tested the signal processing tools, pre-processing, post-processing, model based beam forming and time reversal, on simulated and experimental data. We found that the resultant enhancement of the signals improved our ability to sharpen images and locate as many scattering centers as necessary for our model, including location of the planar interfaces.

We applied the present finite difference codes to carefully posed 3 dimensional problems to avoid the discrete nature of the material interfaces. We supported the development of the discrete surface integral propagation code, but were unable to finish it in time to apply it to our modeling problem. We used the TSAR microwave code as a substitute, but again had to carefully choose the problems approached.

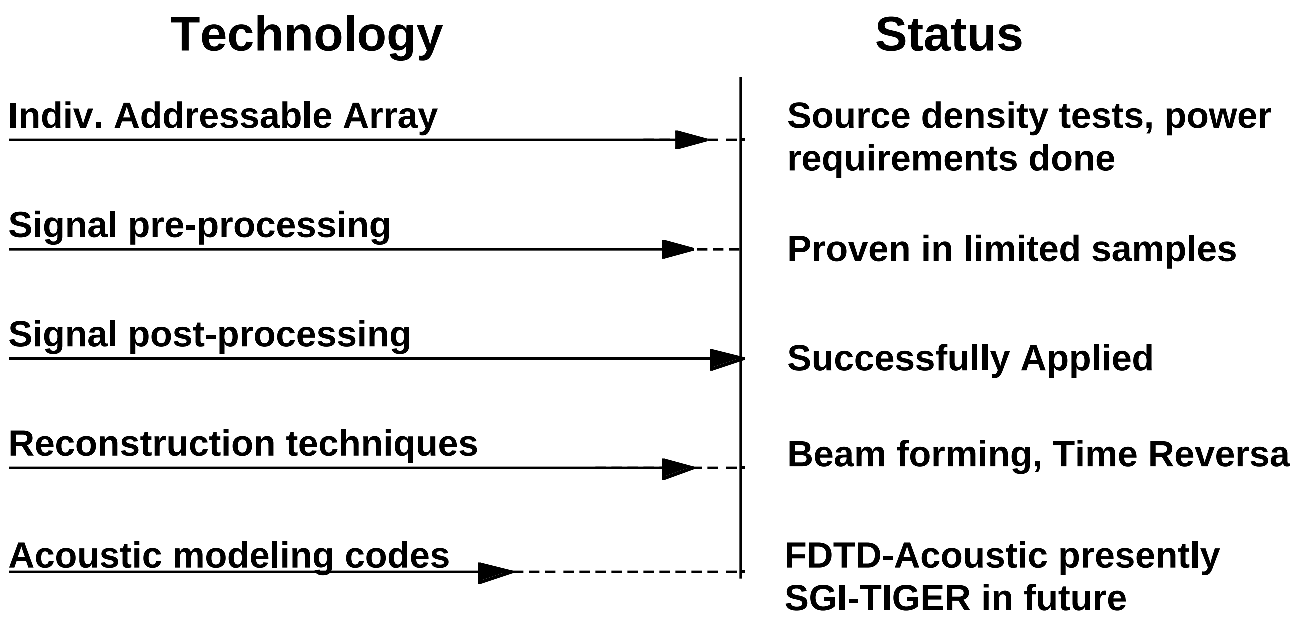

Figure 8. Status of 5 technologies at project end. Individually addressable arrays were not built since true elastodynamic modeling capability was unavailable. Pre-processing of signals was not tested on elastodynamic models, but was tested on simplified models and experimentally.

We have developed a unified method for applying the 5 technologies necessary to create a superior type of 3 dimensional imaging in high contrast and dispersive elastic media. We have determined the configuration of array elements necessary to achieve a resolution of 10s of um resolution. We can match data to a preliminary model and reformulate our model to more closely match each individual situation.

We have found a method to allow tomographic reconstruction without using back propagation calculations, saving computation time. We can improve contrast and resolution by using pre- and post-processing techniques proven in dispersive media.

We have achieved our goals for the first phase of our project, and determined the design parameters of a prototype for use in follow-on studies. We have taken the first step in providing a real time, high resolution, 3 dimensional imaging technology for use in generalized elastic media.